INTRODUCTION
The femoropopliteal (FP) artery is a branch of the femoral artery, the main artery in the upper leg, providing blood to all muscles and superficial tissues in the thigh. It is the largest of the femoral artery branches, composed of the superficial femoral artery (SFA) in the proximal region and popliteal artery (PA) in the distal region which runs below the knee. It is characterised by its tortuous geometry, associating a high atherosclerotic plaque burden with it. However, due to the dynamic forces of the SFA and PA, peripheral stents are reported to have the highest failure rates, predominantly due to bending [1]. Worst case bending can be seen in regions of the SFA/PA behind and just above the knee [2] and this is detrimental to stent patency.

Computational simulations of stent deployment processes and interactions between stent struts and arterial tissue have become increasingly popular in assessing the viability of stent designs as they allow for focus on specific areas of interest and provide a means of comparing in-vivo and in-vitro experiments that can complement each other to give significant conclusions. Results of many studies have been published of complex computational investigations simulating the deployment of stents into blood vessels [3-6]. Results of these investigations allow for the evaluation of stent-artery interactions both during and following stent deployment. Such models can provide much insight into the behaviour of specific stent designs in a physiological surrounding and depict the behaviour of the vessel due to a specific stent deployment. Due to significant developments in the power of super computers and advancements of finite element solvers to date, increasingly sophisticated finite element models can be used to accurately estimate complex stent-artery interactions.

As stenting in the SFA is not only dependent on stent-artery contact but also on the relative movement and contact of the surrounding muscles of the leg, dramatic geometrical changes of the SFA and PA occur leading to high failure rates. The goal of this work was to accurately model stent artery-interaction in the SFA by including the effect of the surrounding muscles during a 90° knee bend using a computational finite element model. Furthermore, an in vitro model of the SFA in a system that is capable of replicating the haemodynamic and biomechanical environment of the SFA using tissue-engineering principles was developed. This involved the augmentation of an existing cardiovascular bioreactor developed for coronary artery simulation [7]. Previous studies have shown the potential of such an in vitro system to evaluate the performance and interaction of stents with the endothelial layer in a controlled environment [8]. The purpose of the model was to determine the effect of bending on cell viability, morphology and orientation in a stented vessel.

EFFECTS OF KNEE FLEXION ON STENTED PERIPHERAL ARTERIES – A COMPUTATIONAL AND IN VITRO STUDY

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METHODS

Computational Model

3D models of the stented SFA, surroundings muscles (adductor longus, rectus femoris, biceps femoris, vastus medialis, vastus lateralis and sartorius) and bones (femur and tibia) were created by importing MRI scan data into MIMICS© software (Materialise NV, Belgium) and constructing anatomically accurate models via manual thresholding in each axial slice. 3-Matic© software (version 5.1, Materialise NV, Belgium) was then used to create finite element mesh representations of the models which were then imported into ABAQUS/Standard (Version 6.10, Simulia, Providence, RI). The stent used for the model was based on the Cordis SMART ™ Nitinol Stent (OD 7mm, length 20mm), created using SolidWorks 3D CAD design software 2008 (Dassault Systèmes Simula Corp, Vélizy, France) imported into ABAQUS/Standard. The analysis involved an initial step where the stent was deployed in the straight SFA and then a second step was initiated where the knee model (consisting of stented artery, bones, muscles) was bent to 90°.

Figure 2. Finite Element Model of Knee Anatomy (a), SFA (b) and SMART stent (c)

In-Vitro Model

Medical-grade silicone (RT-601, Elastosil, Wacker, Germany) was used to produce silicone tubes that were seeded with endothelial cells to produce pseudovessels. Silicone exhibits similar levels of radial distention to those of a native vessel when subjected to physiological levels of pressure. The tubes were coated with human fibronectin (8 mg/ml, Sigmaaldrich, Ireland) and seeded with HUVECs (Clonetics, Cambrex Bio Science, UK). The pseudovessels were rotated for 24 h to allow cell adhesion. The presence of a confluent monolayer was confirmed by H&E staining of a segment of the tube. Following seeding, a SMART ™ Stent (Cordis) was deployed in the pseudovessel. The stented model artery was then transferred into the bioreactor flow loop by attachment in a specially designed support chamber which imposed a 50° bend on the tube (the worst case bend that can be imposed in the tube before kinking occurs). This chamber was then incorporated into the flow loop which mechanically conditioned the stented model artery for 24 hours by allowing physiological levels of pressure and flow (120/80 mmHg and 300ml/min respectively) in an incubator at 37°C.

RESULTS AND DISCUSSION

Computational Model

The results of the finite element model provide great insight into the behavior of the artery due to the surrounding muscles during a knee bend cycle. Furthermore the incorporation of the stent into the analysis allows the resulting stress concentrations in the artery due to the combined in vivo and stent loading to be realized. Worst case locations of stress in the stented artery after knee bending can be easily recognized in the distal SFA region. Furthermore results show that worst case stress locations of the stented portion of the artery occur at the ends of the stent due to the stiffness mismatch between the ends of the stented portion and arterial tissue as shown in Figure 3. Results highlight the negative effects of stiffness mismatch of stented and non-stented portions of the artery post-stenting.

Figure 3. Worst case arterial stress concentrations in the stented region after stenting and in vivo loading from muscles due to 90° knee bend (MPa)

In-Vitro Model

After 24 hours of biomechanical stimulation, model arteries were fixed with methanol (Sigma) and subsequently immersed in Hematoxylin and Eosin (Sigma). The tubes were then cut longitudinally, stent removed and placed on slides for cell morphology evaluation using light microscopy. The angle of cell orientation was determined throughout the length of the model arteries also. In the stented straight control pseudovessel localized endothelial cell denudation was observed along the regions where the stent struts were removed. Cells in between stent strut regions were seen to have altered cell alignment in comparison to the straight unstented control which showed all cells in the stimulated model artery to orient in the longitudinal direction. Comparison of the response and behaviour in a straight model artery with that of the cells in the bent model artery highlights the effect of the bend on cell viability, orientation, morphology and effects of stiffness mismatch in the stented portion.

REFERENCES